APPARATUS AND METHOD FOR BREATH MONITORING

Inventors: Ahmet Karakasoglu
Karl S. Johnson
George D. Hermann

This invention relates to an apparatus and method for breath monitoring.

Attempts have heretofore been made to monitor breathing with various types of devices. Certain of such 5 devices have used thermocouples which depend on heating of the air as one exhales or cooling of the air as one inhales. These have generally been unsatisfactory because of the very long time constants associated with their reaction to breathing. An indirect method is using a 10 pulse oximeter that measures breathing indirectly by measuring oxygen saturation in the blood. Oxygen saturation is a measure of O_2 binding to hemoglobin and typically is denoted by SaO_2. SaO_2 is determined by sending an optical signal in two wavelengths. Hemoglobin 15 is a molecule, binding O_2. Since hemoglobin's light wavelength is a function of its oxygen saturation, oxygen saturation measurements are based upon sending an optical signal and determining the hemoglobin's wavelength according to the reflected signal. Determining breathing 20 disorders by measuring oxygen saturation has several

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important shortcomings. The pulse oximeter is unable to differentiate the saturating gas. In other words, the pulse oximeter readings are the same for either blood which is saturated with O_2 or CO. The SaO_2 response is

5   delayed. Thus for a person experiencing respiratory disorders like sleep apnea, a relatively long period of time such as two hours is required for the oxygen saturation to decrease. A pulse oximeter also does not work reliably when the patient's body is cold or the

10   patient is just coming out of anaesthesia administered for surgery. In addition all these devices have the shortcomings in that they are incapable of accurately measuring air flow volume. There is therefore a need for a new and improved apparatus and method for breath

15   monitoring and for measuring respiratory air flow.

In general, it is the object of the present invention to provide an apparatus and method for providing real-time breath monitoring and respiratory air flow.

Another object of the invention is to provide an

20   apparatus and method of the above character in which the breath is directly monitored.

Another object of the invention is to provide an apparatus and method of the above character in which the breath is monitored by picking up the sound of breathing

25   by the use of a microphone.

Another object of the invention is to provide an apparatus and method of the above character to monitor breathing by sensing vibrations caused by turbulence in the air flow.

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Another object of the invention is to provide an apparatus and method of the above character in which the sound and vibration sensor work accurately with relatively low signal strengths.

5   Another object of the invention is to provide an apparatus and method of the above character in which disordered breathing can be readily ascertained.

Another object of the invention is to provide an apparatus and method of the above character which is 10   operative even though the patient is being administered oxygen.

Another object of the invention is to provide an apparatus and method of the above character which can function in a noisy environment.

15   Another object of the invention is to provide an apparatus and method which can be utilized for monitoring breathing.

Another object of the invention is to provide an apparatus and method of the above character in which the 20   information being generated can be transmitted to a remote location.

Another object of the invention is to provide an apparatus of the above character which is light in weight and easy to use and easy to apply and remove.

25   Additional objects and features of the invention will appear from the following description in which the preferred embodiments are set forth in detail in conjunction with the accompanying drawings.

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Figure 1 is a front elevational view of an apparatus incorporating the present invention mounted on the face of a human being or a patient.

Figure 2 is a side elevational view looking along the line 2-2 of Figure 1 showing the acoustical device as it is positioned on the face of the patient.

Figure 3 is a rear elevational view of the acoustical device shown in Figure 1.

Figure 4 is a cross-sectional view taken along the line 4-4 of the acoustical device shown in Figure 3.

Figure 5 is a block diagram of the electrical circuitry utilized in the apparatus for breath monitoring incorporating the present invention.

In general, the apparatus for breath monitoring monitors respiratory air flow from the nostrils of the nose and/or the mouth of the face of a patient and comprises an acoustical device positioned on the face of the patient in the vicinity of the nose and/or mouth of the patient and having at least one acoustic duct receiving respiratory air flow from the patient. A sensor is provided which is exposed to the acoustic duct and senses turbulence and/or pressure changes in the respiratory air flow in the acoustic duct and provides an electrical output signal. Means is provided for digitally processing the electrical output signal to provide a real-time signal indicative of breathing of the patient.

More in particular, the apparatus 11 for breath monitoring as shown in Figure 1 is depicted as being used on a patient in the form of a human being having a head 16

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having a centrally disposed nose 18 with first and second nostrils 21 and 22 which open downwardly over a mouth 26 formed by upper and lower lips 27 and 28. As shown, the upper lip 27 is positioned below the nostrils 21 and 22 so

5   that there is provided a space 29 to be utilized for a purpose hereinafter described. The head 16 has ears 31 and 32 on opposite sides of the head 16.

The apparatus 11 includes an acoustic device 36 which is formed of a suitable medical grade plastic such as

10 polyurethane, polyvinyl chloride and silicone. The acoustic device 36 is sized so that it is adapted to be mounted on the face 17 of the patient immediately below the lower extremity of the nose 18 and just above the upper lip 27 in the space 29. As shown, the acoustic

15 device 36 can be formed of a single body of material which is relatively flexible so that it can accommodate the contours of the face so that it is comfortable while being worn by the patient.

The body 37 consists of an elongate tubular portion

20 38 which is generally disposed in a horizontal position on the face 17 as hereinafter described and is provided with an acoustic passage 39 extending therethrough. The body 37 is also provided with a tubular portion 41 which extends from the tubular portion 38 and extends at

25 substantially right angles thereto. It is provided with an acoustic passage 42 extending therethrough and is in communication with the acoustic passage 39. The acoustic passage 42 opens through inclined cut or surface 43 provided on the tubular portion 41 which is inclined

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rearwardly and upwardly of the acoustic device 36. The
body 37 is also provided with spaced apart upstanding
tubular portions 46 and 47 which are inclined forwardly
with respect to the vertical axis of the tubular portion
5   41 and are positioned so that they are equidistant
sidewise with respect to the depending tubular portion 41.
The tubular portion 41 has a length, so that when the
acoustic device 36 is positioned on the face of the
patient as hereinafter described, so that the inclined
10 surface 43 is generally in alignment with the mouth 26 and
is exposed to air flow into and out of the mouth 26. The
tubular portion 41 typically can have a length ranging
from 1/2" to 1" and preferably approximately 3/4".
Similarly, the upstanding tubular portions 46 and 47 have
15 a length so that they immediately underlie the nostrils 21
and 22 of the patient when the acoustic device 36 is
positioned on the face of the patient as hereinafter
described. Thus they can have a length ranging from 1/8"
to 1/2" and preferably approximately 1/4". The upstanding
20 tubular portions 46 and 47 have acoustic passages 48 and
49 extending therethrough which are in communication with
the acoustic passage 39 in the elongate tubular portion
38. The acoustic passages 48 and 49 open through inclined
surfaces 51 and 52 provided on the outer or uppermost
25 extremities of the upstanding portions 46 and 47.

An elongate solid but yieldable support member 56 is
formed integral with the body 37 and is formed of the same
material as the body 37 and is generally quite flexible.
As shown particularly in Figure 4, it extends upwardly and

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rearwardly from the elongate tubular portion 38 for a distance of approximately one-half inch. As shown the support member 56 extends at a suitable angle as for example from 10° to 20° and preferably at an angle of

5 approximately 20°. As shown particularly in Figure 3 it can be provided with spaced-apart upwardly facing curved surfaces 57 and 58 with a curved surface 59 extending therebetween.

An adapter 66 is mounted on tubular portion 38.

10 Alternatively as shown, this adapter 66 can be formed integral with the tubular portion 38. This adapter 66 is provided with an acoustic chamber 67 in communication with the acoustic passage 39. A sensor 71 is mounted in the acoustic chamber 66 and has a surface which is exposed to

15 the acoustic passage 39 and is provided for sensing turbulence and/or pressure changes in the acoustical chamber 67 in communication with the acoustic passage 39 and provides an electrical signal on electrical conductors 72 connected to the sensor 71. The sensor 71 can be a

20 vibration sensor or a microphone. The sensor 71 since it is exposed to the acoustic passage 39 will sense the vibrations and pressure changes generated by turbulent air flow within the acoustic passage 39.

In order to accommodate the needs of hospital

25 environments and even in certain home environments, it is desirable to be able to provide the patient with oxygen in the event that this is required by the patent. Thus the acoustic device 36 can be used in much the same way as a

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nasal cannula conventionally utilized for delivering oxygen to a patient.

In order to increase the accuracy of the apparatus for breath monitoring, it is desirable to provide another sensor 76 for suppressing ambient sounds or noises in the vicinity of the patient. For this purpose, another tee-shaped adapter 77 is mounted on the other end of the tubular portion 38 opposite the end on which the adapter 66 is mounted. It can be a separate part or it can be formed integral with the tubular portion 38. It provides an acoustic chamber 78 which is in communication with the acoustic passage 39 and which opens into a centrally disposed leg 79. The sensor 76 is mounted in the leg 79 and is in the form of a microphone and is utilized for sensing ambient sounds and/or noise in the vicinity of the patient. The sensor 76 is connected to conductors 81 extending from the leg 79.

When it is desired that the apparatus 11 be versatile enough so that it can administer oxygen to the patient at the same time that breath monitoring is taking place, an oxygen tube set 86 is used. It consists of first and second tubes 87 and 88 which are connected to a wye adapter 91. The wye adapter 91 is provided with a side arm 92 which is connected to a suitable source of oxygen shown as "O_2 SOURCE" as for example an oxygen supply from the hospital piping or alternatively from a compressed oxygen container through an appropriate regulator apparatus (not shown). Oxygen is thereby supplied to both of the tubes 87 and 88 and which extend through a slip

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ring 96 slidably mounted on the tubes 87 and 88. The
tubes 87 and 88 branch out in a wye in opposite directions
and extend over the ears 31 and 32 of the patient and are
connected to the adapters 66 and 77. If necessary, to
5 additionally secure the tubes 87 and 88 in close contact
on the face, pieces 90 of adhesive tape can be used as
shown.

Since the sensor 71 is positioned within the adapter
66, oxygen will only be supplied through the other
10 flexible tube 88 into the adapter 77 and thence into the
upstanding tubular portions 46 and 47 as hereinafter
described. The sensor 71 when oxygen is being delivered
to the patient will also sense the constant oxygen flow
which will also act as a disturbance signal contributing
15 to the turbulence and pressure changes in the chamber 67.
In such an arrangement, the conductor 72 will extend
through the flow passage 89 and the tube 87. The
conductors 81 will extend through the flow passage 91 of
the tube 88 through the wye adapter 91 and into electrical
20 coupling 96 which mates with another electrical coupling
97 to provide a disconnect between the oxygen tubing and
electrical cable 98 which is connected to the coupling 97.
The other end of the cable 98 is connected to a plug-in
jack 99. The jack 99 as shown is plugged into a console
25 101 which is a sleep-disorder breathing detection device
utilizing advanced audio digital signal processing
technology as disclosed in co-pending application Serial
No. 08/472,441, filed June 7, 1995, now Patent No.
5,797,852. As described therein, the device is typically

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intended for unattended home studies of obstructive sleep apnea which has been designed to be simple to use, low cost and a non-invasive device with screening system accuracy. Typically the device calculates and stores some
5 of the vital statistical information: time and duration of apneas and hypopneas, respiratory sound intensity, duration of patient snoring and respiratory disturbance index.

      The console 101 houses a digital signal processor
10 chip as well as memory and analog input preamplification circuitry which is powered from a conventional source of power such as 110 volts AC by use of a plug-in adapter 102 connected by a power cord 103 to the console 101. The console is provided with an inclined front panel 106
15 having provided thereon a start button 107, a stop button 108 and a back lit display 111. Typically the information provided by the display 111 includes a bar graph whose length is proportional to a snoring sound intensity and the following:

20   D - sleep study date

             T - 24 hour running system clock reference

           ST - sleep study start time

           ET - sleep study end time

           TH - total breath monitoring study time minus
25   study pauses (if any)

          DBE - cumulative disordered breathing events
                during study

          AHI - apnea hypopnea index; average number of
                DBE's per hour

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When the patient is ready to sleep, the start button 107 is pressed. The next morning the patient presses the stop button 108 to end the breath monitoring study.

5   The electrical circuitry which is utilized in the apparatus and method for breath monitoring is shown in block diagram form in Figure 5. As shown in Figure 5, the sensor 71 is a vibration sensor whereas the alternative sensor 71a is a microphone. Let it be assumed that the acoustic device 36 has been mounted on the face of the

10 patient in the manner shown in Figure 1 in which the oxygen tubes 87 and 88 are looped over the ears 31 and 32 and that the acoustic device has been positioned so that it is between the upper lip and nose as shown with the bottom extremity of the tubular portion 41 being in front

15 of the mouth 26 between the upper and lower lips 27 and 28 of the patient and at the same time that the support member 56 rests against the upper lip of the patient in such a manner so that the upward extending tubular portions or canula 46 and 47 immediately underlie the

20 nostrils 21 and 22 of the patient as shown in Figures 1 and 2.

Let it also be assumed that the oxygen tube set harness 86 is connected to an oxygen source and is also connected to the control console 101. The breathing

25 monitoring study is now ready to be undertaken. By utilizing a vibration sensor as shown in Figure 5, the vibration sensor 71 should be relatively immune to ambient sounds and should only be sensing the turbulence and pressure changes occurring in the acoustic passage 39

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created by the patient breathing through the nostrils 21 and 22 of the nose 18 and/or through the mouth 26 with the air passing to and from the mouth being sensed by the extending tubular portion 41 and the air flow into and out

5 of the nostrils being sensed by the canula or upwardly extending portions 46 and 47 of the body 37 of the acoustic device 36.

In order to obtain a greater accuracy in the breath monitoring, and particularly where noisy environments are

10 encountered, an ambient sound sensor in the form of a microphone 76 is utilized to pick up the sound. The ambient sound sensor typically is in close vicinity to the face of the patient and the vibration sensor 71 as for example as being located on the other end of the acoustic

15 device 36 as shown in Figure 3. When oxygen is being supplied to the patient, the constant oxygen flow is measured in a conventional manner. The flow of oxygen will affect the turbulence and pressure changes occurring within the acoustic passage 39. In order to remove this

20 constant flow oxygen signal, the ambient sound sensor or microphone 76 placed outside of the respiration area of the patient collects signals generated by the constant oxygen flow and also by ambient sounds and produces a signal which is proportional to the oxygen flow which is

25 used to minimize the effect of the oxygen flow in the respiratory air flow to the patient. In connection with the foregoing it should be appreciated that the ambient sound sensor 76 should be isolated from the vibration sensor 71 in order to minimize acoustical coupling.

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In the present invention by placing the sensor 71 in an acoustical chamber 67 which is in communication with acoustical ducts 39, 42, 48 and 49, the sensor 71 is measuring pressure changes and turbulence in the air flow

5 as well as measuring sound. In addition, it has been found very desirable to at the same time that air flow measurements are being made to also measure respiratory sounds of the same patient or in other words ambient sounds in the vicinity of the nose and mouth of the

10 patient by the use of the second sensor or microphone 76. Thus breathing sounds and snoring by the patient can be picked up by this second microphone 76 as well as other ambient noise. This microphone 76 acts as a respiratory sound sensor which is dedicated to sense the sound

15 intensity in dB. It is placed outside of the air stream or streams created by the breathing of the patient in order to avoid the turbulence and vibrations generated by direct exposure to air flow. Thus, the second sensor is a proximity microphone predominantly receiving respiratory

20 sounds but also receiving ambient noise.

The turbulence sensor microphone 71 and the ambient sound sensor 76 should be carefully positioned so that the acoustical and mechanical coupling between them is minimized. Thus the signal and noise components at the

25 output of the vibratory sensor 71 are uncorrelated but the noise at the output of the respiratory sound sensor 76 is correlated with the noise component of the sensor 71 when a microphone is used.

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The electrical circuitry in the console 13 is that which is disclosed in co-pending application Serial No. 08/472,441 filed on June 7, 1995, now Patent No. 5,797,852. The circuitry pre-amplifies and A to D

5 converts the signals received from the sensor 71 and the ambient sensor 76 in the form of dedicated sensors. These sensors supply analog signals which are amplified by analog preamplifiers and fed to A/D converters to an active noise canceler which contains an adaptive filter

10 that uses the ambient sound sensor 76 as a reference to produce an estimate of the ambient noise which is subtracted from the primary sensor 71 output as represented in block 131. The output of the active noise canceler is used to adjust the tap weights in the adaptive

15 filter. The adaptive filter minimizes the mean-square value of the overall output thereby causing the overall output to be the best estimate of the desired signal in the minimum-mean-square sense.

Efficient ambient noise suppression in the present

20 invention is accomplished by developing a model for the acoustical transfer function of the space interconnecting the two sensors. The model employed in the present invention is in the form of a transversal filter consisting of a set of delay line elements each of which

25 is represented by one sampling period of delay and a corresponding set of adjustable coefficients. At the sampling time instant, k, the available signal consists of a set of samples

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$$u(k),\ u(k-1),\ u(k-2)\ ...u(k-n). \tag{1}$$

These samples are convolved with a corresponding set of adjustable tap weights

$$\tilde{\omega}_0,\ \tilde{\omega}_1,\ \tilde{\omega}_2,\ ...,\ \tilde{\omega}_n$$

to produce an output signal sequence, $y(n)$ with the signal
5   collected by the ambient noise sensor 76 being $d(n)$. The
filter output $y(n)$ is compared with the $d(n)$ to produce an
estimation error $e(n)$ which is used by the adaptive
algorithm whereby it controls the corrections applied to
the individual tap weights. This procedure is continued
10 until the noise estimation error $e(n)$ becomes sufficiently
small in order to encounter the possible directions of
arrival to the sensors. To keep the estimation algorithm
casual during its operation, an internal time delay of d
is added to the reference input signal. The time duration
15 of d is selected according to the sampling rate and the
assumed distance between the two microphones.

     For a sampling rate of 8000 Hz and a distance of 1 cm
between the two microphones, the sound propagation delay
is about

$$T = \frac{d}{c} = \frac{1\,cm}{320\,m/s} = 31.25\ microseconds \tag{2}$$

20   Should the acoustical properties of the testing
environment change, the adaptive algorithm used to adjust
the coefficients has the added task of continually
tracking the variations of the system. For this reason a

normalized adaptation step size has been used to improve the convergence rate.

The electrical circuitry in the console 101 also includes a band-pass filter bank consisting of several narrow band-pass filters as described in co-pending application Serial No. 08/472,441 filed June 7, 1995, now Patent No. 5,797,852, which information is incorporated herein by reference. The software performs an appropriate selection of a frequency band in order to improve the sound digital quality, i.e., reduce any interference of ambient noise. The task of this software module is to determine a frequency band where the signal-to-noise ratio for a given turbulence or vibration is maximum. In general, the useful portion of the sensor frequency spectrum is within the frequency band of 100 Hz to 1500 Hz. The filtering section consists of ten filters each having a band width of 300 Hz and having cutoff frequencies from 200 Hz to 1500 Hz.

When the signal from the active noise canceler reaches the band-pass filter it is applied to each of the ten filters and corresponding outputs are computed. Since the input signals are known, the filter that yields the best signal-to-noise ratio is selected and used for each successive three-minute period. At the end of each three-minute period, the selection process is repeated in order to compensate for any possible variations in the vibration or turbulence characteristics.

The filters in the console 101 are realized by using Butterworth's technique. These filters have a flat

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response in the pass-band and are monotonic overall. The filter is of autoregressive-moving average type, having both poles in zero so that as sharp as possible attenuation rate is obtained with a minimum number of taps. The 3rd order Butterworth type filter output $y(k)$, can be described by

$$y(k) = a_1 y(k-1) + a_2 y(k-2) + \ldots + a_6 y(k-6) + b_0 u(k) + b_1 u(k-1) + \ldots + b_6 u(k-6). \quad (3)$$

For this application, all the a_i and b_i coefficients are calculated by using the Matlab routine

[numerator, denominator] = Butter(filter order, cutoff frequencies).

This ambient noise and artifact suppression hereinbefore described is set forth in box 132 of Figure 5. This step is followed by a step of sound/air flow conversion utilizing EVA waveforms as set forth in box 133 in Figure 5. If sensors 71 and 76 are both microphones, the sensor 71a receives respiration plus O_2 flow signals whereas the second sensor 76 receives predominantly O_2 flow. The ambient sounds are received by both microphones 71a and 76. Therefore a suitable subtraction of sensor 76 signals from sensor 71a signals improves the quality of the breathing or respiration signals being obtained. As pointed out previously since the sensors 71a and 76 are not co-located, the analysis heretofore given must take into account the acoustical transfer function between the sensors 71a and 76. Thus the signals received from the box 132 are converted to air flow signals utilizing the EVA transformation shown in block 133.

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In order to determine whether the patient's breathing
is disordered, the amount of air volume inhaled and
exhaled needs to be known. The microphone signal which is
available after the steps shown in box 132 have been
5   performed still are not very informative as to air flow
volume. In other words even though the sound signal
contains the required air flow volume information the
sound signal at this point does not directly provide it.
For this reason and in accordance with the present
10 invention the waveform is generated and a scoring
methodology is utilized for this waveform that closely
mimics the results of the core of a conventional scoring
mechanism using a standard polysomnograph. If desired,
the waveforms generated can be displayed on a CRT to
15 permit viewing or visual inspection by a skilled
technician. The signal after it has been subjected to the
steps shown in box 132 is subjected to a process where it
is converted to an estimated volume of air flow (EVA)
waveform that represents an estimate of its acoustical
20 energy. This conversion procedure, which is explained in
application Serial No. 08/472,441 filed June 7, 1995, now
U.S. Patent No. 5,797,852, is performed digitally in a
discrete time domain and is somewhat similar to a high
order integration process in a continuous-time domain. In
25 accordance with the present invention, the following
prediction sequence is utilized for generating the
estimated volume of air flow waveforms that are expected
to closely follow the actual air volume that is inhaled or

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exhaled by the patient. The software code implements the following equation for this purpose.

$$\sigma(k) = \alpha_1\ \sigma(k1) + \alpha_2\ \sigma(k2) + \alpha_3\ \sigma(k3) + \ldots + \alpha_m\ \sigma(k/m) + \beta|x(k)| \qquad (4)$$

where α and β are user defined parameters that change the shape of the waveforms. It is found that

5    $\alpha_1 = 0.995$, $\alpha_i = 0.0$ for $i=2, 3, \ldots,$ m and $\beta = 0.2$ were suitable choices or obtaining a waveform that closely resembles air flow waveforms that are obtained by using the standard polysomnograph. To obtain a condition for stability, the z-transform of this equation yields

$$(z - \alpha)\ \sigma(z)\ = \beta z\ X(z). \qquad (5)$$

10   Then,

$$\sigma(z)\ = \beta\ [z/(z - a)]\ X(z). \qquad (6)$$

a selection of

$$0 < \alpha < 1$$

is sufficient to obtain stability for the algorithm.

The next which follows the step shown in box 133 are
15   steps shown in box 134 for the classification of disordered breathing events.

As hereinbefore pointed out, once a respiration sound has been sensed by the respiratory sound sensor 71 or 71a it must be processed to ensure that it is indeed a breath
20   pattern. After that has been ascertained, its characteristics can be analyzed to detect disordered breathing. Since these parameters are mostly patient-dependent, a set of self-tuning algorithms that

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characterize a given breath signature are utilized. This has been accomplished by utilizing intermediate variables in the form of the estimated volume of air flow (EVA) waveforms hereinbefore described and thresholds to obtain

5 a very close scoring correlation with that of the standard polysomnograph. At the same time, if desired, the frequency of breathing can be ascertained.

As explained in co-pending application Serial No. 08/472,441 filed on June 7, 1995, now U.S. Patent No.

10 5,797,852, a pattern classifier is utilized for operating on the EVA waveforms which is a rule-based decision making unit. It operates on the EVA waveforms for the determination of disordered breathing events by employing a set of time-varying coefficients that are designed to

15 adapt to a given breathing pattern. The related set of parameters are adjusted to score an event when the cessation of a less or less than normal air flow levels have been encountered for a duration of ten seconds or longer. The less than normal definition that has been

20 followed applies to an air flow reduction by 50% relative to nominal levels. For a classification, a time-varying nominal value is first assigned by detecting peaks of the EVA waveform. Based on this value, the apnea threshold for the determination of less than normal breathing levels

25 is computed. The remainder of the EVA waveform is then monitored to detect any events taking place.

After the classifications called for in the block 134 have been accomplished, the information is stored in a memory of a suitable type and the sleep-disordered

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breathing events occurring can be displayed in an LCD shown by the box 137 much in the same manner as described in co-pending application Serial No. 08/472,441 filed on June 7, 1995, now U.S. Patent No. 5,797,852, or
5 alternatively on a CRT for visual observation.

Also in accordance with the present invention, once a signal has been received after the pre-amplification A/D data conversion as represented by the box 131, additional information can be obtained without any hardware changes.
10 Thus it is possible to readily provide information on sound level intensity. This is accomplished by utilizing the information supplied by the ambient sound sensor 76 which picks up the sound field which may include respiratory sounds. In accordance with the present
15 invention, this signal is sampled at 5.525 kHz rate. A plain logarithmic conversion provides sound levels in decibels (dB). A plain logarithmic conversion is identified as dB_c. Where human perception is involved, especially below 1000 Hz, another decibel scale has been
20 developed identified as dB_A. Thus in accordance with the present invention, the console 101 is utilized to calculate dB_A by digitally weighting the frequencies below 1000 Hz so that the characteristics closely match that of published dB_A curves.
25 For a direct conversion from the signal after the box 131, dB_c is computed with the collected sensor signal being denoted by $x(t)$ and its logarithmic correspondent $y(t)$ is

$$y(t) = 20 \log_{10} \{s(t)/S_{ref}\}; \tag{7}$$

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$$and \ s(t) = f \ [\Sigma \ x(t)]$$

where $f(p)$ is a function that performs a frequency waiting over a collected time window samples and s is the reference.

5   The dB$_A$ conversion process is performed by a well-parameterized software code to provide flexibility to address different needs. In connection with the present invention, the console 13 makes dB$_A$ measurements five times per second and stores. After collecting one minute long

10   dB$_A$ data, minimum peak dB, maximum peak dB and one-minute windowed average (two 300 samples) dB values are transferred to the processor memory. At the end of the study, along with other disordered breathing information and statistics, the dB level information containing

15   minimum, maximum and average value measurements at every minute are stored into the non-volatile random access memory (NVRAM). The dB measurement is calibrated by using an accurate meter from the ambient noise floor (approximately 40 dB$_A$) to 110 dB$_A$ at every five dB$_A$

20   intervals.

From the foregoing description from the use and operation of the apparatus in performing the method of the present invention, the following sequence events are utilized.

25   1.   Turbulence and vibration in respiratory air flow
              sounds as well as ambient sounds are picked up
              by the dedicated sensors

         2.   The signal quality is improved and noisy
              components are removed by a filter bank

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3. The purified air flow sound signal is converted to an air flow curve called EVA

4. Based on the EVA waveforms, disordered breathing events are identified

5. Apnea and hypopnea events are classified

6. The purified respiratory sound signal is used to calculate the sound intensity level (dB)

7. The display is updated

8. The data are stored in an internal non-volatile random access memory (NVRAM) chip

9. Display on a CRT

10. Initiate alarm in the event of abnormal breathing

From the foregoing it can be seen that an apparatus and method has been provided for breath monitoring which provides a more accurate means for recognizing disordered breathing patterns and for providing respiratory sound intensity. The apparatus has built-in operational intelligence and has been designed to make the apparatus have a simple appearance and so that it can function reliably in a home environment and at the same time be user friendly. Thus time and duration of apneas and hypopneas can be readily classified into events. At the same time sound intensity levels can be displayed at every three minutes. Sufficient storage capacity is provided so that four ten-hour breath monitoring studies can be retained. A Windows®-based software code makes it possible to retrieve, display and print the collected data on a personal computer platform. A summary report and

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event log are provided for reviewing the sleep data quoted.

Although the apparatus and method have been principally described in conjunction with real-time
5 studies for disordered breathing in adults, it should be appreciated that the apparatus and method can also be utilized for monitoring small infants for monitoring their breathing non-invasively. The apparatus and method can also be utilized for monitoring the performance of
10 athletes and how they react under stress. By utilizing radio frequency telemetry as hereinbefore described, the athletes can be monitored even when they are on the athletic field or doing workouts. Conventional wireless methods may be utilized for conveying the information to
15 the desired site. In addition, the apparatus and method can be utilized for monitoring the breathing of patients undergoing anesthesia and other medical procedures.

When oxygen is being administered to the patient at the same time that breathing of the patient is being
20 monitored, the active noise cancellation requires the implementation of a notch filter type response to suppress tonal natured noise due to constant oxygen flow to the patient. This makes it necessary to model the transfer function for the acoustic path between the two sensors so
25 that ambient sounds are canceled.

From the foregoing it can be seen that the apparatus and method of the present invention makes it possible to ascertain relative air flow levels continuously, ascertaining breathing frequency in breaths per minute to

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establish the breathing form and to alarm when a breath
pattern is found to be disordered.

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